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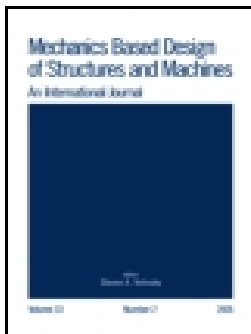
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# **Mechanical Design of an Affordable Adaptive Gravity Balanced Orthosis for Upper Limb Stroke Rehabilitation**

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## **Abstract**

In this paper a novel design of a non-powered orthosis for upper limb stroke rehabilitation is reported. Its design exploits the gravity balancing theory. Designed for home based use, it is the first affordable, passive design to incorporate an assistive level that can be adaptively varied within a closed-loop control scheme. This allows the device to be integrated with a dual robotic and electrical stimulation control scheme, to thereby enable full exploitation of the motor relearning principles which underpin both robotic therapy and Functional Electrical Stimulation (FES) based stroke rehabilitation. This embeds the potential for more effective treatment. The

paper focuses on the mechanical design of the non-powered orthosis, providing detailed design, dynamic analysis and evaluation.

**Keywords:** FES, gravity balancing theory, multibody dynamics, passive orthosis, stroke rehabilitation

## INTRODUCTION

Worldwide 12.6 million people live with moderate to severe disability following stroke, and this number is increasing. Due to the neuroplasticity of the brain, the function of the damaged cells can be transferred to surrounding areas of the motor cortex (Egglestone et al. 2009). Hence impaired subjects can potentially regain lost motor function with the help of a tailored rehabilitation programme involving training of motor tasks, with feedback (e.g. proprioceptive, haptic, visual) used to forge new motor connections. The therapeutic effectiveness of these therapies strongly depends on their frequency, intensity, and the amount of voluntary effort supplied by the patient. Unfortunately only 5% of patients receiving conventional therapy, involving manual assistance supplied by a physiotherapist, recover useful movement (Barreca et al. 2003).

In the field of non-conventional rehabilitation, new upper-limb technologies have the potential to provide intensive and motivating therapy. Moreover, there is evidence that non-conventional therapies offer more effective treatment than conventional ones (Egglestone et al. 2009; Mehrholz et al. 2009; Loureiro et al. 2011). Rehabilitation robotics is a leading non-conventional therapy and the employment of robots has been shown to be very effective for patient recovery, however their design, control and use are complex, often expensive, hence they are not suitable as home-based devices. On the other hand, non-powered orthoses can offer an

alternative as low cost support devices. They are usually made of a mechanism combined with spring systems or counterweights that balance a mass on their extremity, for instance the user's arm, thus the patient can power the orthosis without carrying the weight of their own arm.

The main drawback of these mechanical systems is that they are not as flexible and effective as rehabilitation robots, and they can be used only by patients with sufficient residual muscle strength to power the orthosis and perform the rehabilitation therapy. To overcome this drawback, in the research reported in this paper, a non-powered orthosis will be combined with Functional Electrical Stimulation (FES) (Lynch and Popovic 2008; Brend, Freeman, and French 2012). FES is another approach to therapy which facilitates movement by applying short electrical pulses to contract muscles in the same way as achieved by the unimpaired central nervous system. FES is often combined with a mechanical device to provide controlled stimulation, especially during reaching tasks. Hence, using a biomechanical representation of the human arm within the FES control scheme, it is possible to adaptively adjust the assistance provided to each patient, encouraging maximum voluntary effort and therefore maximising effectiveness of treatment (Freeman et al. 2009, 2012).

This paper proposes a novel design of an orthosis that is suitable to combine with FES, thereby unlocking the potential of effective rehabilitation in an affordable package. The design of the orthosis is based on the gravity balancing theory of mechanisms, by combining a parallelogram linkage with zero-free length springs (Agrawal, Gardner, and Pledgie 1999; Herder and Tuijthof 2000) for lightweight design. The discussion in this paper focuses on the design process, from the static gravity balancing analysis of the linkage to the 3D CAD model and the preliminary dynamic analysis of the arm support.

# EXISTING PROTOTYPES AND GRAVITY BALANCING

## THEORY

The most relevant non-powered orthoses to meet the aim of this research are the Wilmington Robotic Exoskeleton (WREX) (Rahman et al. 2006) and Armon Orthosis (Herder 2005; Herder et al. 2006; Mastenbroek et al. 2007). These are both based on a mechanical design that makes use of spring systems to achieve gravity balancing.

WREX is an exoskeleton design-based body-powered orthosis that provides gravity support to allow patients to move their arm with very little effort (Rahman et al. 2006). This result was achieved by employing elastic bands rather than using counterweights, as in many other devices, giving rise to a lighter and more compact mechanism.

The Armon Orthosis is a non-powered mechanism that can be mounted on a wheelchair for the purpose of assisting activities of daily living (ADLs) (Herder 2005; Herder et al. 2006; Mastenbroek et al. 2007). The orthosis does not resemble an exoskeleton, as in the case of WREX. Indeed, the joints of the mechanism do not have the function of emulating the anatomical human arm joints, but they are arranged to support the weight of the user's arm, when combined with a spring-cable-pulley system located at the lower extreme of the orthosis. The simplicity and efficiency of the Armon Orthosis is due to its mechanical design that is strongly based on gravity balancing theory for more details refer to (Herder 2001).

At this stage it is important to introduce the concept of statically balanced systems. They are systems that stay in a static equilibrium state throughout their range of motion in the absence of friction. Mechanical devices designed to exploit this property present many advantages in terms of improved mechanical performance, including the ability to support the arm over a large

workspace, while simultaneously presenting minimal inertial characteristics to the patient. This is critical in the design of a body worn orthosis, where it is important to reduce the effect of the weight of the supported limb, and the inertia of the mechanical system. A statically balanced system can be constructed by exploiting at least two conservative forces arranged in such a way as to provide stability. This result can be achieved by means of two potential energy storage devices, for instance springs and masses. Moreover these system components must be coupled in such a way that their energy characteristics, as a function of the d.o.f., add up to a constant value.

For the purpose of this research it was decided to use the *zero-free length* springs model in the design of the balancing system, since they have the fundamental characteristic that the force is proportional to the actual length of the spring, rather than the sum of the initial length and the elongation as in normal springs (Herder 2001). This means that the employment of this type of springs simplifies the analysis and the design of spring mechanisms, leading to linear mechanical models. It also simplifies the subsequent hybrid control scheme.

## MECHANICAL DESIGN

In this section the mechanical design of the orthosis will be discussed in detail. The static gravity balancing analysis of the mechanism is developed to derive the most suitable mechanical solution for the adjustment system. Then all the features and the mechanical subsystems of the preliminary 3D CAD model of the arm support are illustrated in detail.

### Kinematic Chain and Gravity Balancing

The novelty of the proposed approach lies in combining FES and a mechanical support to produce an affordable system which delivers adaptable and personalisable assistance during functional task completion. The mechanism utilized to achieve this objective is shown in **Figure**

1. Its hybrid kinematic chain embeds the favourable mechanical features of both parallel and serial linkages (Carbone, Cannella, and Angeles 2011). Moreover, the orthosis differs from that of typical exoskeletons, since it provides gravity balancing to the patient's arm by acting only on a small area of the forearm, identified by the Combined Center of Mass (CCM), making use of a parallelogram linkage and zero-free length springs. This mechanical solution has been used in several existing devices, but for this particular application the objective is to make it adjustable and lightweight, whilst maintaining an affordable design.

The model comprises a planar two d.o.f. kinematic chain made of a parallelogram linkage combined with two *zero-free length springs*, each balances a single d.o.f. of the mechanism.

The length of the links is  $L$  and  $r$ . Moreover  $r$  and  $a$  are also the geometrical parameters of the springs. The moving end of each of the two springs of stiffness  $k_1$  and  $k_2$  is connected to point  $A$  and point  $B$ , respectively, whereas the fixed end is connected to the point  $O$ . The mass  $m$  of the user's arm is assumed to be attached to point  $F$  on the distal link, which corresponds to the CCM. Now it is possible to analyse the gravity balance behaviour of the mechanism, where the mass of the moving links is balanced by the springs (Herder 2001). The mass of each of the links is considered as lumped at their center of mass. Focusing on the  $i$ -th spring, it is possible to decompose its force  $k_i l_i$ , where  $l_i$  is the spring elongation (equivalent to length in the case of *zero-free length springs*), into a component along the link length  $r$ ,  $k_i r$ , and a component along  $a$ ,  $k_i a$ , on  $\overline{OC}$ .

Due to the linearity of the model, it is possible to use the superposition principle for the gravity compensation analysis. Considering **Figure 2(a)** the first step is to fix spring  $k_1$  and perform the moment equilibrium about pivot  $C$ . The only moment contribution of spring  $k_2$  is



that due to component,  $k_2 a$ . The masses  $m$  and  $m_4$  can be shifted to point  $D$ , and the mass  $m_3$  parallel to link 2, from  $CM_3$  to  $CM'_3$ , as shown by the arrows in **Figure 2(a)**. This procedure is possible because after shifting the masses, the respective shape of their trajectories remains identical and so their potential energy differs only by a constant value. Hence the moment equilibrium equation about pivot C, when spring  $k_1$  is fixed, can be written as

$$(m + m_4)gL + m_3 gr_{m3} + m_1 gr_{m1} = k_2 ar, \quad (1)$$

where  $r_{m1}$  and  $r_{m3}$  are the respective distance of the centre of mass  $CM_1$  and  $CM'_3$ , from the pivot C.

The second step is to release spring  $k_1$  and fix spring  $k_2$ , in order to perform the moment equilibrium about the point  $D$ , as shown in **Figure 2(b)**. The only moment contribution of spring  $k_1$  is due to its component  $k_1 a$ , while the masses  $m_3$  and  $m_2$  can be shifted to point  $E$  and  $CM'_2$ , respectively, as shown in **Figure 2(b)**, for the same reason as above. Hence the moment equilibrium equation is

$$mgL + m_4 gr_{m4} - m_2 gr_{m2} - m_3 gr = k_1 ar, \quad (2)$$

where  $r_{m2}$  and  $r_{m4}$  are the respective distance of the centre of mass  $CM'_2$  and  $CM_4$  from point  $D$ . Both (1) and (2) are linear and can be combined into the linear system of equations

$$\begin{cases} (m + m_4)gL + m_3 gr_{m3} + m_1 gr_{m1} = k_2 ar \\ mgL + m_4 gr_{m4} - m_2 gr_{m2} - m_3 gr = k_1 ar. \end{cases} \quad (3)$$

These contain all the geometrical and inertial elements of the mechanism and of the springs, that can be varied in order to adjust the gravity balancing behaviour of the device.

Analyses reveal that the variation of the linkage parameters is complex to achieve and may require a heavy mechanism directly acting on it. Hence, the implementation of mechanisms for the adjustment of springs stiffness could be an alternative to the previous ones, but it usually requires mechanical systems that lack of precision and accuracy (Herder 2001). To avoid these difficult and cumbersome mechanical solutions,  $a$  was chosen as the adjustment parameter, and this leads to the design of an adjustment system that can be integrated within the lower part of the mechanism, without acting directly upon its links, but varying the position of the pivot  $O$ , relative to  $C$ .

### 3D CAD Model: Components and Materials

The 3D CAD model of the orthosis consists of an engineering implementation of the gravity balanced linkage analysed in Section 3.1, combined with an interface connection, an adjustment system and a structural support. The 3D CAD model shown in **Figure 3** has been developed using SolidWorks and has in total five d.o.f. divided as follows: Two d.o.f. for the interface connection, two d.o.f. for the parallelogram linkage and one d.o.f. for the vertical axis of rotation of the mechanism (the fifth revolute joint in **Figure 3**). The additional d.o.f. permits more dexterity to the subject's forearm whilst performing rehabilitation tasks. The range of motion of the orthosis is such that it allows the subject to perform typical stroke rehabilitation tasks, such as reaching. Indeed it is expected that the end user will mainly perform tasks which do not involve a wide abduction rotation of the shoulder. Moreover, the arm support must be able to provide gravity balancing to a variety of different subjects, each of whom will have different anthropomorphic characteristics, and hence different arm weights. This justifies the design requirement of including an adjustable balancing mechanism, while the overall manufacturing cost is kept low by the choice of materials and off-the-shelf components. The interface

connection between the user's arm and the support is to be made of a folded aluminium sheet with two revolute joints, one with a horizontal axis and another with the axis orthogonal to the sheet.

The CCM of the user's arm has to be placed on it, so these two d.o.f. enable more forearm dexterity during use of the device. The interface connection will be later improved by covering it with molded foam and straps to ensure comfortable, safe and stable contact with the user's forearm. Deep groove ball bearings were selected to be mounted in the joints of the device in order to reduce friction and so improve the mechanical performance of the overall system. Moreover, less friction translates into less effort for the patient when using the device during treatment.

The four links of the parallelogram mechanism are to be made also of aluminium, with characteristic dimensions of  $L$  and  $r$ , as defined in **Figure 1**. The length  $L$  was chosen as 320 mm in accordance with the design studies conducted for the Armon Orthosis prototype (Herder et al. 2006; Mastenbroek et al. 2007), while  $r$  was chosen as 50 mm in order to accommodate the balancing mechanism and the adjustment system. All the sheets and plates used for the arm support were selected to be made of steel. This choice will reduce the overall manufacturing cost of the device.

The lower part of the linkage is connected to the springs via a *string-pulley* arrangement (Ottaviano, Castelli, and Cannella 2008), shown in **Figure 4**. This solution allows the gravity balancing system to have a linear mechanical behaviour, by using normal extension springs, rather than *zero-free length springs*, reducing the manufacturing cost of the device. Examining in detail one of the two *string-pulley* arrangements (one for each d.o.f) in **Figure 4**, it can be seen how one end of the string is connected to the upper hook of the spring, then wrapped around the

first pulley placed at the bottom part of the linkage. The same string is wrapped around two more pulleys placed on the movable folded metal sheet. This sheet is movable since it is connected to the adjustment system, which varies parameter  $a$ .

The adjustment system, shown in **Figure 5**, is made of a ball screw connected to a brushless motor, and a guide. This solution is designed so that it can be later integrated with a control system that allows the balancing of the device to be adjusted according to the value of the supported mass. At a later stage a hybrid FES controller will be implemented to automatically vary the support given to the user's arm based on their rehabilitation progression. Ball screw systems were chosen for their high accuracy and precision in positioning, and lower friction.

The adjustment system allows variation of the parameter  $a$  within the range of 0 mm, when the mechanism is folded, to 50 mm, when  $a = r$  and the system balances the maximum value of the mass. This maximum mass of 3 kg was chosen according to studies conducted for the design of the Armon orthosis (Herder et al. 2006; Mastenbroek et al. 2007). The *string-pulley* balancing system was designed to perform this adjustment avoiding interference with the links of the mechanism. The lower hooks of both springs are connected to the plate, to which the fixed part of the ball screw system is attached. This specific plate is connected to a revolute joint with a vertical axis that allows the horizontal adduction of the user's shoulder. This means the whole mechanism is able to rotate around the vertical axis. This joint is bigger than the others since it has to carry all the weight of the mechanism, and fit the brushless motor in its upper flange, reducing the overall volume and the inertia about the vertical axis of the system.

The revolute joint is then connected to a frame, which comprises welded steel squares. The purpose of the frame is to carry the load of the whole mechanical system and to provide a stiff and steady support during its use. Moreover, four casters with brakes are installed at the

bottom part of the frame, in order to make it easier to move and locate the orthosis in different environments or rooms.

An example of how the orthosis is used by a subject is shown In **Figure 6**, where a 3D CAD model of the human arm has been combined with the one of the orthosis for illustrative purposes. It is possible to notice how the connection area between the interface connection and the human arm involves only the forearm.

## DYNAMIC ANALYSIS OF THE ORTHOSIS

In this section the dynamic behaviour of the orthosis is analysed. The first study involves the linkage. A dynamic analysis was performed in order to evaluate the performance of the linkage in the case of using either bushings or bearings in the joints. Then, a dynamic analysis of the 3D CAD model of the orthosis was performed in order to assess the gravity balancing property under dynamic conditions.

### Dynamic Analysis of the Linkage

A dynamic analysis of the linkage of the device is now conducted to confirm appropriate performance. The model is based on the planar linkage of **Figure 1**, where an extra d.o.f. has been added to describe the rotation around the vertical axis of the overall parallelogram linkage. Its detailed view is shown in **Figure 7**. The general dynamic equation for robotic systems is

$$\mathbf{B}(\mathbf{q})\ddot{\mathbf{q}} + \mathbf{C}(\mathbf{q}, \dot{\mathbf{q}})\dot{\mathbf{q}} + \mathbf{F}_v\dot{\mathbf{q}} + \mathbf{F}_s\text{sgn}(\dot{\mathbf{q}}) + \mathbf{g}(\mathbf{q}) = \boldsymbol{\tau} - \mathbf{J}^T(\mathbf{q})\mathbf{h}_e, \quad (4)$$

where,  $\mathbf{q}$ ,  $\dot{\mathbf{q}}$  and  $\ddot{\mathbf{q}}$  are the  $(n \times 1)$  generalized coordinate, velocity and acceleration vectors, respectively,  $\mathbf{B}(\mathbf{q})$  is the  $(n \times n)$  inertia matrix,  $\mathbf{C}(\mathbf{q}, \dot{\mathbf{q}})$  is the  $(n \times n)$  Coriolis matrix,  $\boldsymbol{\tau}$

is the  $(n \times 1)$  vector of the actuation torque at the actuated joints,  $\mathbf{F}_v \dot{\mathbf{q}}$  and  $\mathbf{F}_s \text{sgn}(\dot{\mathbf{q}})$  are the  $(n \times 1)$  vectors of the viscous and static friction torques, respectively,  $\mathbf{h}_e$  is the  $(n \times 1)$  vector of the contact forces and moments acting on the end-effector of the manipulator and  $\mathbf{J}$  is the  $(m \times n)$  *Jacobian matrix* of the mechanism.

In this specific case the number  $(n)$  of active joints of the mechanism is 3, and the generalized coordinate vector is  $[\vartheta_0 \quad \vartheta_1' \quad \vartheta_1'']^T$ . The overall dynamic analysis of the system has been performed using ADAMS (MSC Software). The static gravity balance analysis of Section 3.1 was used to calculate the stiffness values of both springs,  $k_1 = 3869\text{N/mm}$  and  $k_2 = 4454\text{N/mm}$ , which were then added to the ADAMS model.

Then, the problem was analysed for the case where joints are equipped with either bushings or bearings, which can be considered as corresponding to the case of either high or low friction, respectively. In the case of bushings, using a steel-brass contact model, the static friction coefficient is  $\mu_s = 0.35$ , and the dynamic friction coefficient is  $\mu_d = 0.19$ . For bearings, they are  $\mu_s = 0.0024$  and  $\mu_d = 0.0012$ , respectively. The dynamic analysis has been made for three different values of mass  $m$ : 1, 2 and 3 kg, acting on the end-effector, while the whole mechanism is under the action of gravity. This analysis is performed to qualitatively show the possible differences in the dynamic performance of the linkage with two different models of joint couplings.

The results in **Figure 8** show the position of the center of mass of the end effector during a simulation of 10 s, for the two friction cases. In the case of bearings (**Figure 8(a), (c) and (e)**) the system tends to become less oscillatory in response to initial conditions when the mass is increased. The reason is that the stiffness of the springs are fixed parameters that are derived

from the static gravity balance analysis, and the chosen values are the ones which corresponds to  $m = 3\text{kg}$ . Furthermore, the results in the plots take into account inertia effects. In the case with bushings (**Figure 8(b), (d) and (f)**), the system is less oscillatory for all values of  $m$ , but this means that the device will not be suitable for weaker patients, since they might not be able to perform a rehabilitation task due to the high friction in the joints of the mechanism. Thus the solution with bearings is chosen as the optimal one, but the plots of the dynamic simulation of the linkage show that a controller must be added in order to vary the parameter  $a$  of the springs, so that it will adapt its dynamic behaviour to different masses, making it more stable and suitable for use by different patients.

### 3D CAD Model Dynamic Simulation

The dynamic simulation of the 3D CAD model is carried out using SolidWorks with the Motion Analysis module. The study is performed analysing the displacement of the centre of mass of the interface connection, when a specific payload is carried by it. The results are presented in terms of coordinate components for the displacement of the centre of mass. The aim is to verify that the maximum displacement is less than 10 mm, which represents a tolerable end-point tracking error for a stroke patient while performing daily living tasks (Brend, Freeman, and French 2012). For the sake of simplicity the two d.o.f. of the interface connection are locked during the simulations. The device can adapt its balancing behaviour for a range of different payloads. Four simulations were performed, where the values of the payload  $m$  and the distance  $a$  have been calculated according to the linear relationship (3). The input parameters for the simulation are reported in **Table 1**.

The stiffness of the two springs is calculated by solving (3) which guarantees the gravity balancing condition. The calculated spring stiffness values are  $k_1 = 4048\text{N/mm}$  and  $k_2 = 5035\text{N/mm}$ , which are kept constant during all four simulations. Since the joints have been designed using bearings, the static friction is  $\mu_s = 0.0024$  and the dynamic friction is  $\mu_d = 0.0012$ . The simulations are run for a duration of 20 s. The coordinates of the centre of mass of the interface connection are shown in **Figures 9–11**.

The results show how the dynamic behaviour of the device improves when the payload increases. The reason is that the stiffness of the springs is a fixed parameter that has been calculated for the case of the maximum payload,  $m = 3\text{ kg}$ . Moreover the oscillatory trend of the centre of mass is due to the lack of damping and to the fact that the stiffness of the springs and the parameter  $a$  were set using a static gravity balancing analysis. These confirm that the device is able to hold a payload and carry its own with a minor level of oscillation, but indicate that a controller is required in order to overcome this drawback, as was the case for the dynamic analysis results of the linkage in Section 4.1. It should be also taken into account that on the real prototype, these oscillations will be mitigated by the additional natural damping introduced by the patient's arm. The only case in which the oscillation amplitude is higher than the limits of 10 mm is the case with zero mass. However, this specific case does not represent a problem, since a payload  $m = 0\text{ kg}$  either implies that no payload is present on the interface connection, hence no patient is using the device, or that patients have regained their skills and they do not need any support from the device. Hence the dynamic analysis confirms the support and gravity balancing objectives of the design.

## CONCLUSION AND FUTURE WORK



This paper proposes the mechanical design of the first device to combine an adaptable gravity balanced orthosis with FES, in order to facilitate affordable and effective upper limb stroke rehabilitation. The details concerning the gravity balancing analysis of the linkage and the design embodiment are presented, with a detailed explanation of the 3D CAD model. Then a dynamic analysis of the linkage and of the 3D CAD model have been carried out, supported by results that confirm the design objectives of the device. This represents a framework for the manufacturing of the device and its design validation.

Future work will comprise manufacturing and testing of the prototype, and will initially involve simulation and analysis carried out with the 3D CAD model. The system will then be integrated with FES using a hybrid controller which contains both a dynamic model of the support mechanism and a dynamic model of the user's arm. Mechanical support and applied FES will both be selected to minimise an objective function involving tracking error and control effort, enabling optimal real-time adjustment in the levels of mechanical and FES assistance. The integrated device will be later tested and evaluated with end-users.

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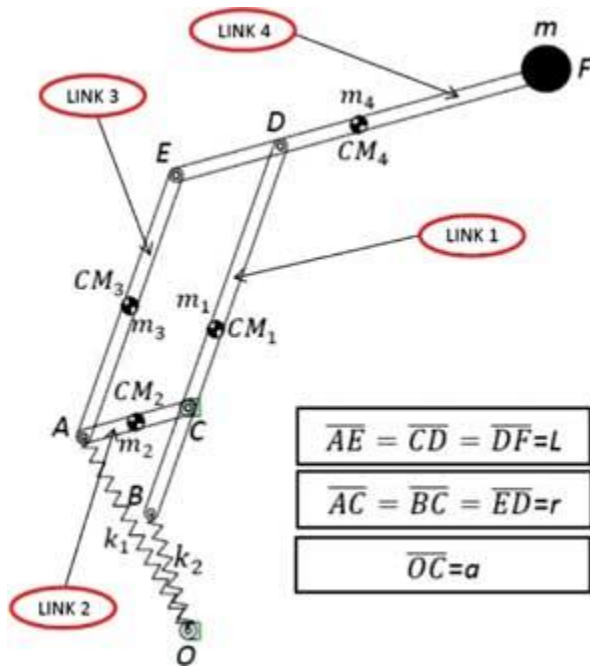
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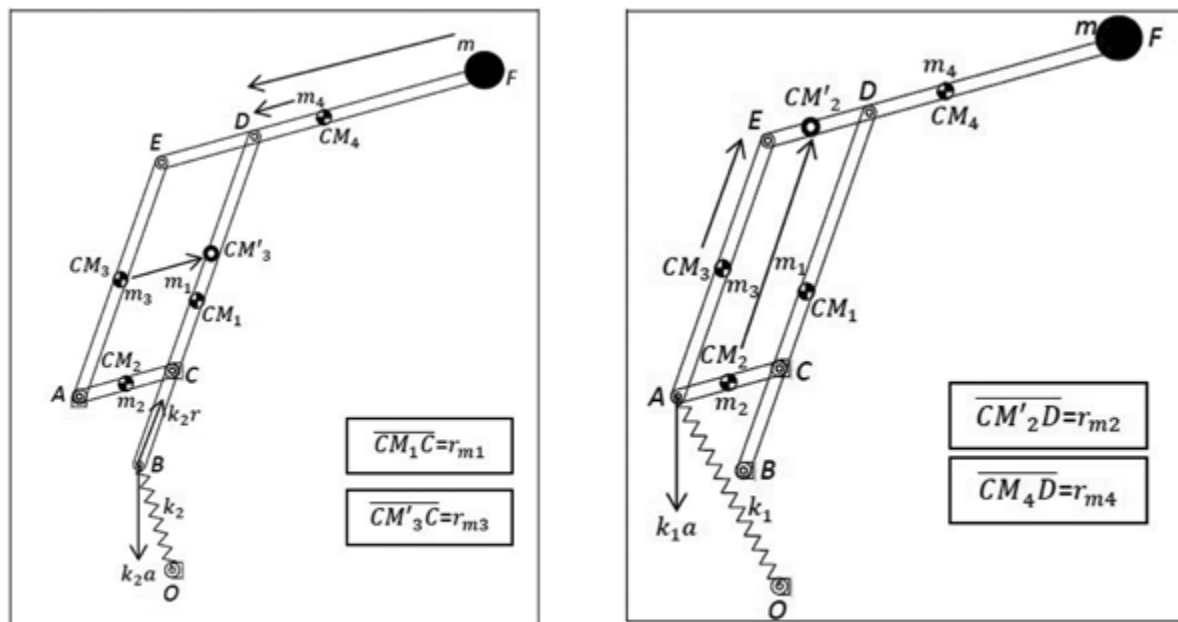
**Table 1.** Values of payload  $m$  and distance  $a$  for the simulations.

$m$ [kg]	$a$ [mm]
0	3.5
1	19
2	34.5
3	50

**Figure 1.** Kinematic chain of the arm support.



**Figure 2.** Moment equilibrium analysis. (a) Moment equilibrium analysis when spring  $k_1$  is fixed. (b) Moment equilibrium analysis when spring  $k_2$  is fixed.

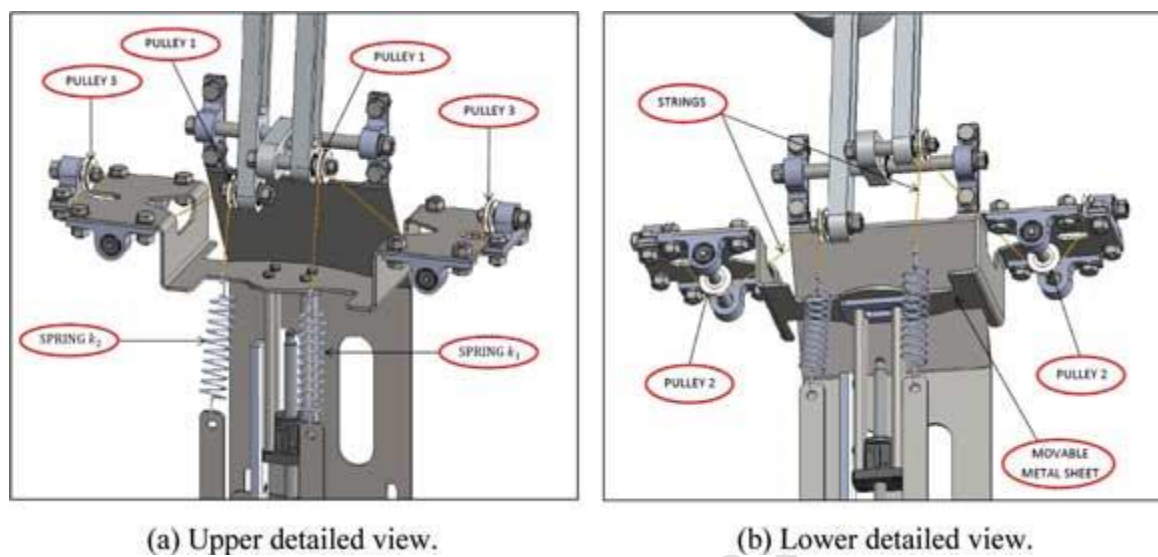


(a) Moment equilibrium analysis when spring  $k_1$  is fixed. (b) Moment equilibrium analysis when spring  $k_2$  is fixed.

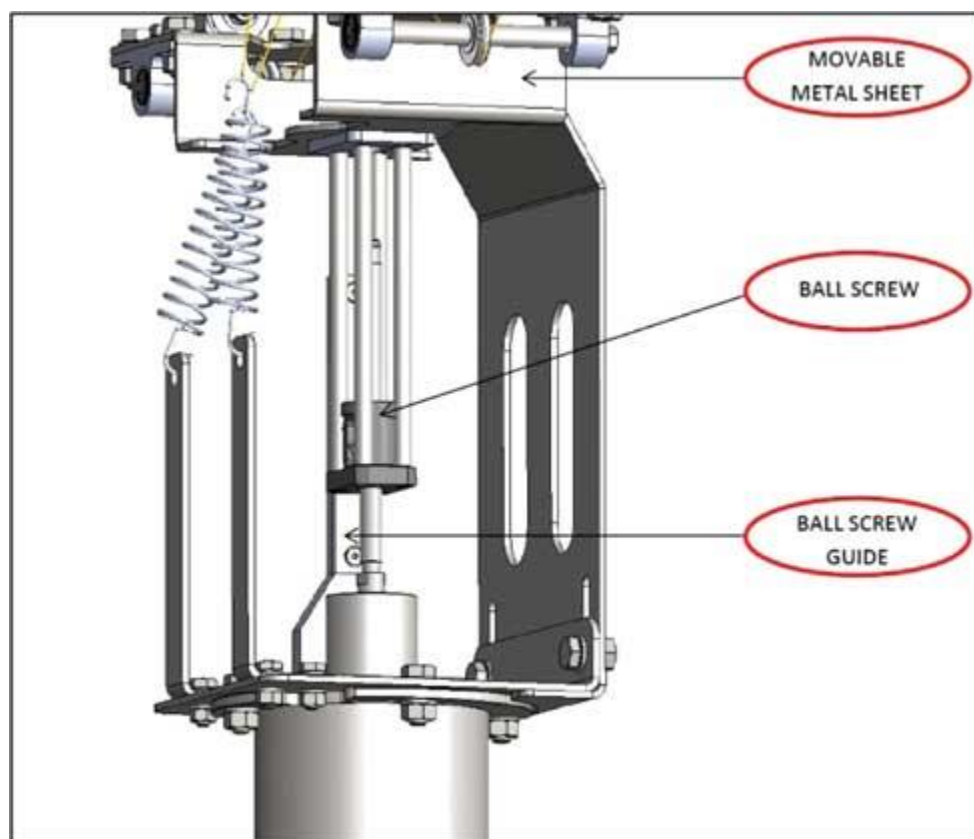
**Figure 3.** 3D CAD model showing subcomponent labels.



**Figure 4.** Detailed views of the gravity balancing mechanism. (a) Upper detailed view. (b) Lower detailed view.

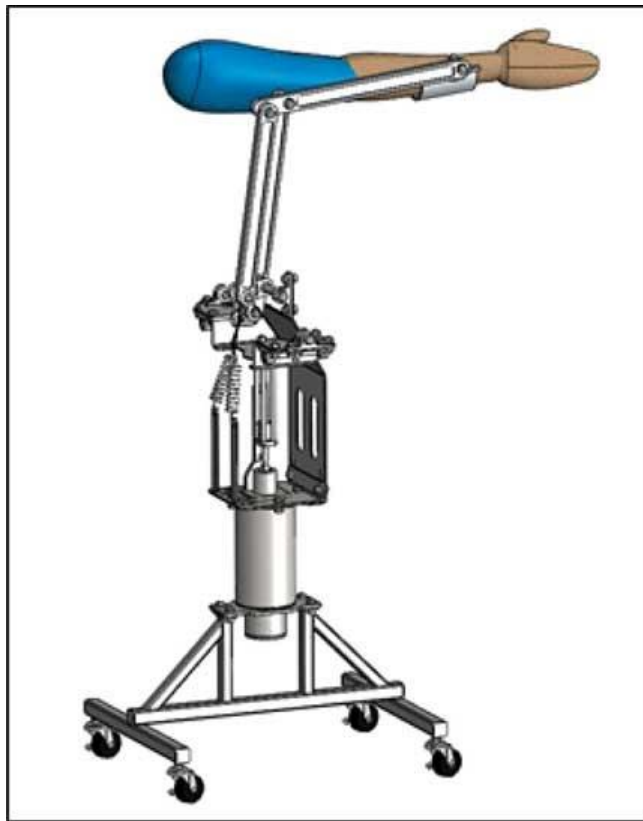


**Figure 5.** Detailed view of the adjustment system.

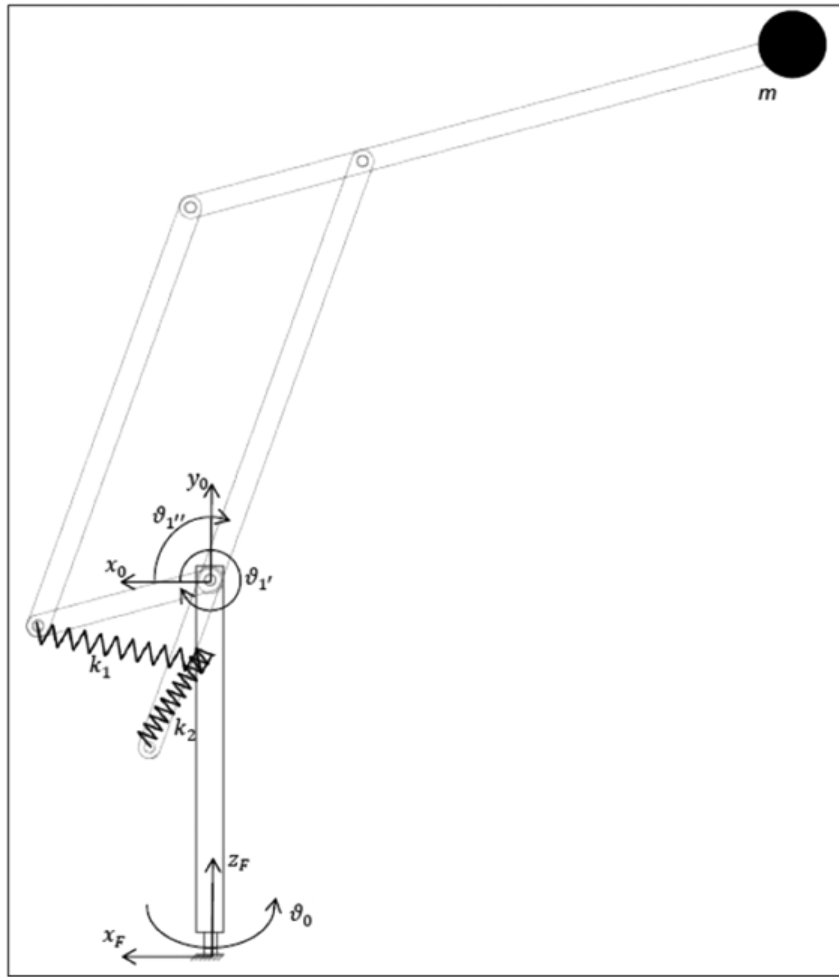




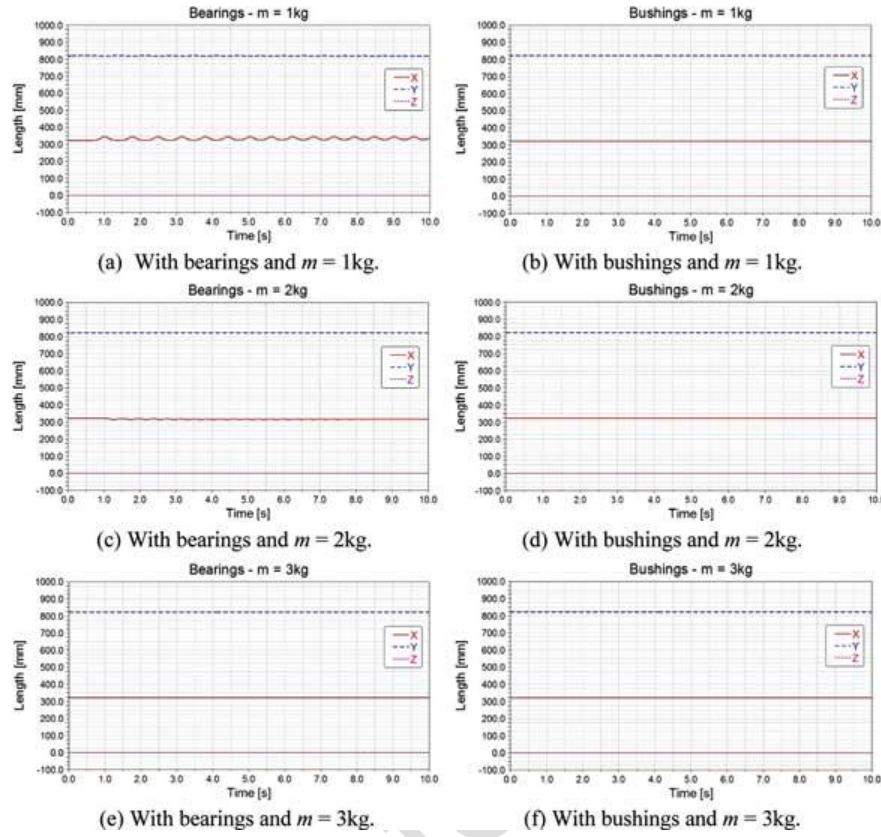
**Figure 6.** 3D CAD model of the human arm and the orthosis.



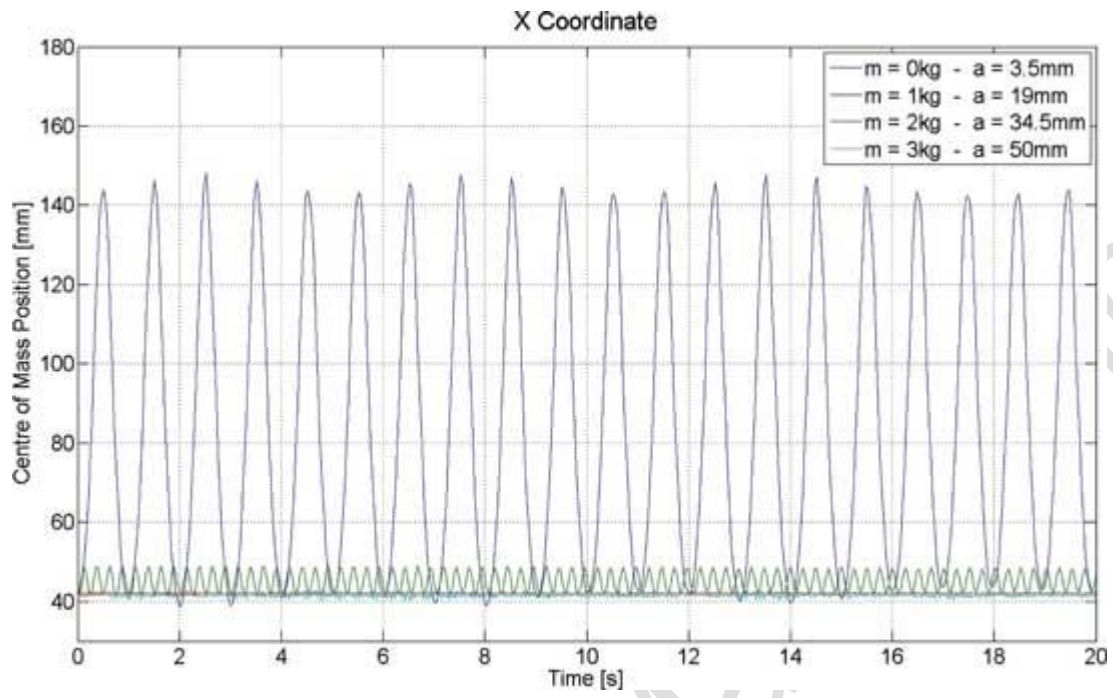
**Figure 7.** ADAMS model for the dynamic analysis.



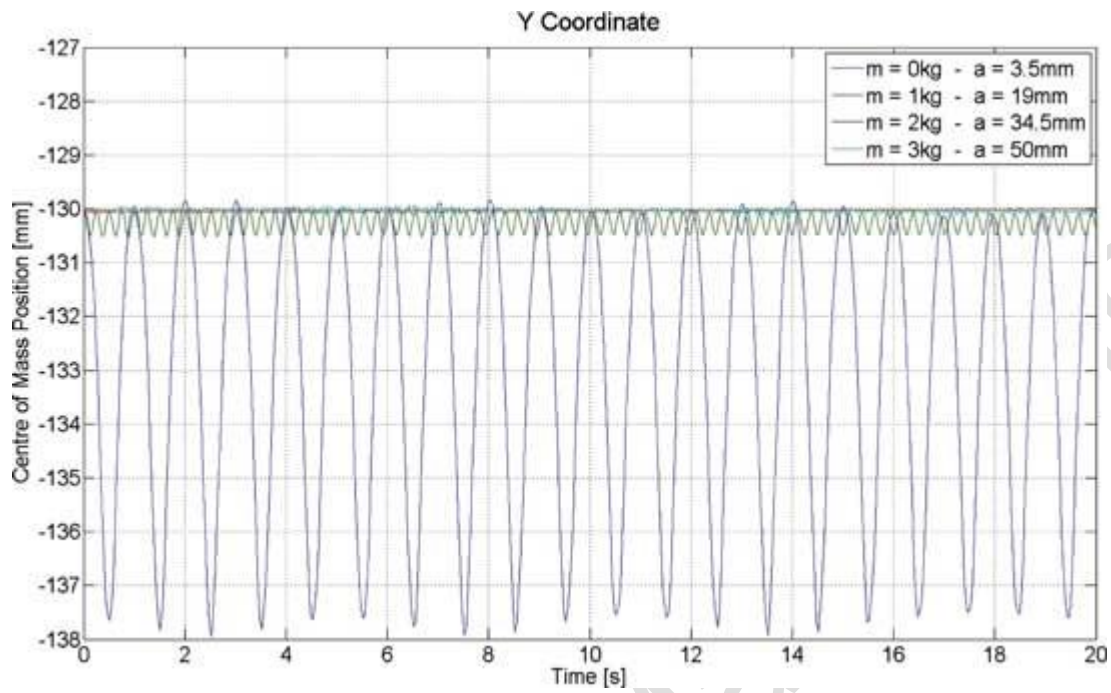
**Figure 8.** Position of center of mass. (a) With bearings and  $m = 1$  kg. (b) With bushings and  $m = 1$  kg. (c) With bearings and  $m = 2$  kg. (d) With bushings and  $m = 2$  kg. (e) With bearings and  $m = 3$  kg. (f) With bushings and  $m = 3$  kg.



**Figure 9.** X coordinate of the position of the centre of mass.



**Figure 10.** Y coordinate of the position of the centre of mass.



**Figure 11.** Z coordinate of the position of the centre of mass.

